NTNU Norwegian University of Science and Technology
Faculty of Medicine > Department of Circulation and Imaging

Basic ultrasound for clinicians

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This web page is intended as an introduction to basic ultrasound physics and technology for clinicians without technical or mathematical background. A basic knowledge of the physical principles underlying ultrasound, will give a better understanding of the practical limitations in ultrasound, and the technical solutions used to solve the problems. This will give a clearer picture of the reasons for the problems and artefacts. No mathematical background will be necessary.

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Ultrasound

Ultrasound is simply sound waves, like audible sound. Although some physical properties is dependent on the frequency, the basic principles are the same. Sound consists of waves of compression and decompression of the transmitting medium (e.g. air or water), travelling at a fixed velocity. Sound is an example of a longitudinal wave oscillating back and forth in the direction the sound wave travels, thus consisting of successive zones of compression and rarefaction. Transverse waves are oscillations in the transverse direction of the propagation. (For instance surface waves on water or electromagnetic radiation.)

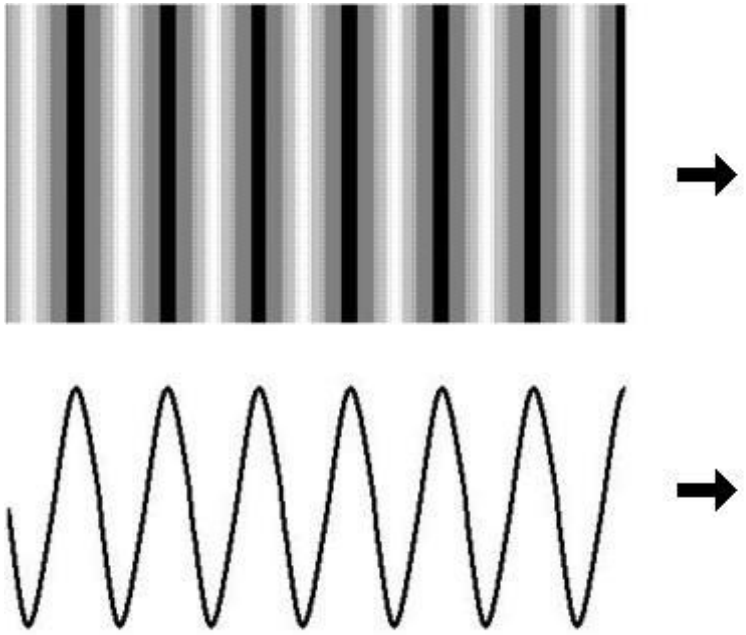


Fig. 1. Schematic illustration of a longitudinal compression wave (top) and transverse wave (bottom). The bottom figure can also represent the pressure amplitude of the sound wave.

The audible sound frequencies are below 15 000 to 20 000 Hz, while diagnostic ultrasound is in the range of 1 - 12 MHz. Audible sound travels around corners, we can hear sounds around a corner (sound diffraction). With higher frequencies the sound tend to move more in straight lines like electromagnetic beams, and will be reflected like light beams. They will be reflected by much smaller objects (because of sorter wavelengths), and does not propagate easily in gaseous media.

At higher frequencies the ultrasound behaves more like electromagnetic radiation.

The wavelength λ is inversely related to the frequency f by the sound velocity c:

$$c = \lambda f$$

Meaning that the velocity equals the wavelength times the number of oscillations per second, and thus:

$$\lambda = \frac{c}{f}$$

The sound velocity i a given material is constant (at a given temperature), but varies in different materials (117):

Material	Velocity (m/s)

Air	330
Water	1497
Metal	3000 - 6000
Fat	1440
Blood	1570
Soft tissue	1540

Ultrasound is generated by piezoelectric crystals that vibrates when compressed and decompressed by an alternating current applied across the crystal, the same crystals can act as receivers of reflected ultrasound, the vibrations induced by the ultrasound pulse .

Imaging by ultrasound

Basically, all ultrasound imaging is performed by emitting a pulse, which is partly reflected from a boundary between two tissue structures, and partially transmitted (fig. 2). The reflection depends on the difference in impedance of the two tissues.

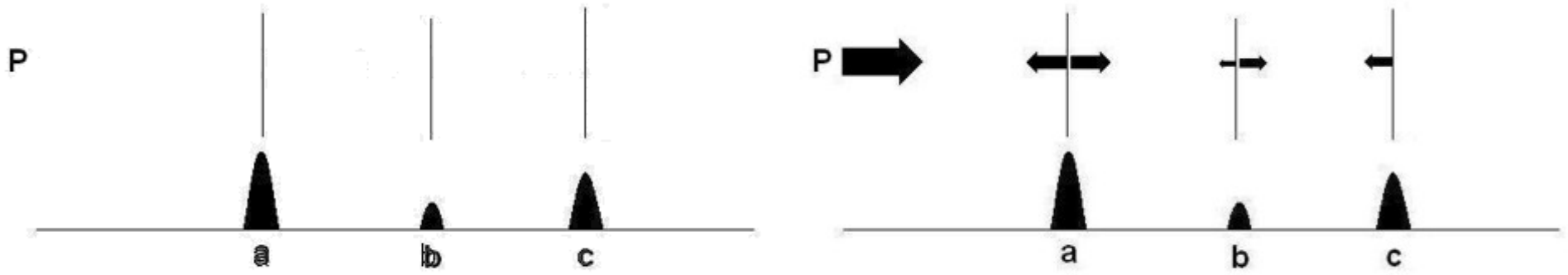


Fig. 2. Schematic illustration of the reflection of an ultrasound pulse emitted from the probe P, being reflected at a, b and c. Part of the pulse energy is transmitted from the scatterer a, the rest is transmitted, part from b and the rest from c. When the pulse returns to P, the reflected pulse gives information of two measurements: The amplitude of the reflected signal, and the time it takes returning, which is dependent on the distance from the probe(twice the time the sound uses to travel the distance between the transmitter and the reflector, as the sound travels back and forth). The amount of energy being reflected from each point is given in the diagram as the amplitude. When this is measured, the scatterer is displayed with amplitude and position. Thus, the incoming pulse a a is the full amplitude of P. At b, the incoming (incident) pulse is the pulse transmitted through a. At c, the incident pulse is the transmitted pulse from b. (In bot cases minus further attenuation in the interval.)

The ratio of the amplitude (energy) of the reflected pulse and the incident is called the **reflection coefficient**. The ratio of the amplitude of the incident pulse and the transmitted pulse is called the **transmission coefficient**. Both are dependent on the differences in acoustic impedance of the two materials. The acoustic impedance of a medium is the speed of sound in the material × the density:

$$Z = c \times \rho$$

The reflecting structures does not only reflect directly back to the transmitter, but scatters the ultrasound in more directions. Thus, the reflecting structures are usually termed scatterers. The time lag, τ, between emitting and receiving a pulse is the time it takes for sound to travel the distance to the scatterer and back, i.e. twice the range, r, to the scatterer at the speed of sound, c, in the tissue. Thus:

$$r = \frac{c\,\tau}{2}$$

The received energy at a certain time, i.e. from a certain depth, can be displayed as energy amplitude, A-mode. The amplitude can also be displayed as the brightness of the certain point representing the scatterer, in a B.mode plot. And if some of the scatterers are moving, the motion curve can be traced by letting the B-mode image sweep across a screen or paper as illustrated in fig. 3. This is called the M-mode (Motion).

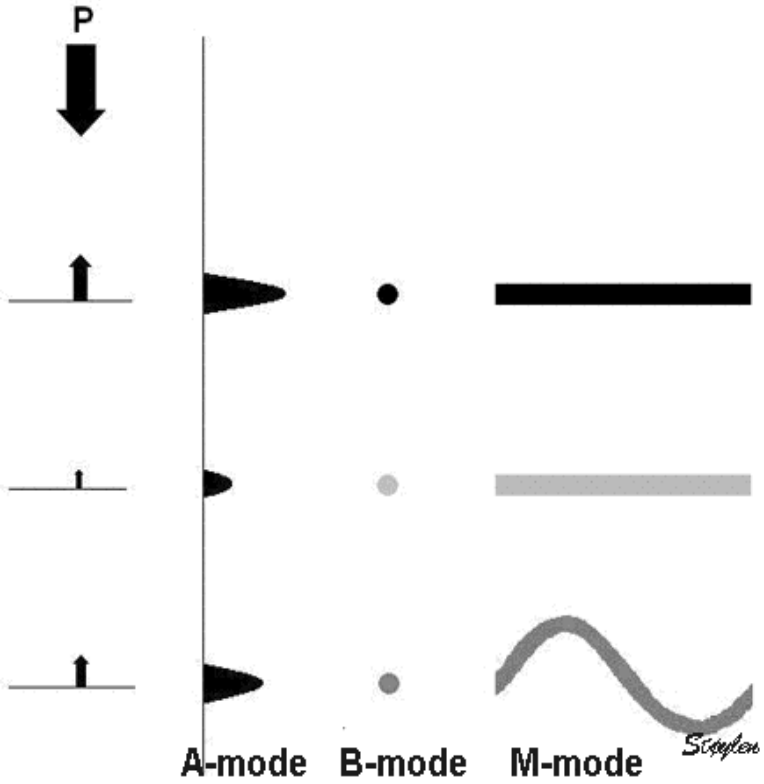


Fig. 3. Ultrasound sent into the depth displayed in three different modes. A-mode (Amplitude) shows the depth and the reflected energy from each scatterer. B-mode (Brightness) shows the energy as the brightness (in this case the higher energy is shown darker, against a light background) of the point. The bottom scatterer is moving. If the depth is shown in a time plot, the motion is seen as a curve, (and horizontal lines for the non moving scatterers) in a M-mode plot (Motion).

M-mode

The M-mode was the first ultrasound modality to record display moving echoes from the heart (118), and thus the motion could be interpreted in terms of myocardial and valvular function. The M-modes were originally recorded without access to 2-dimensional images.

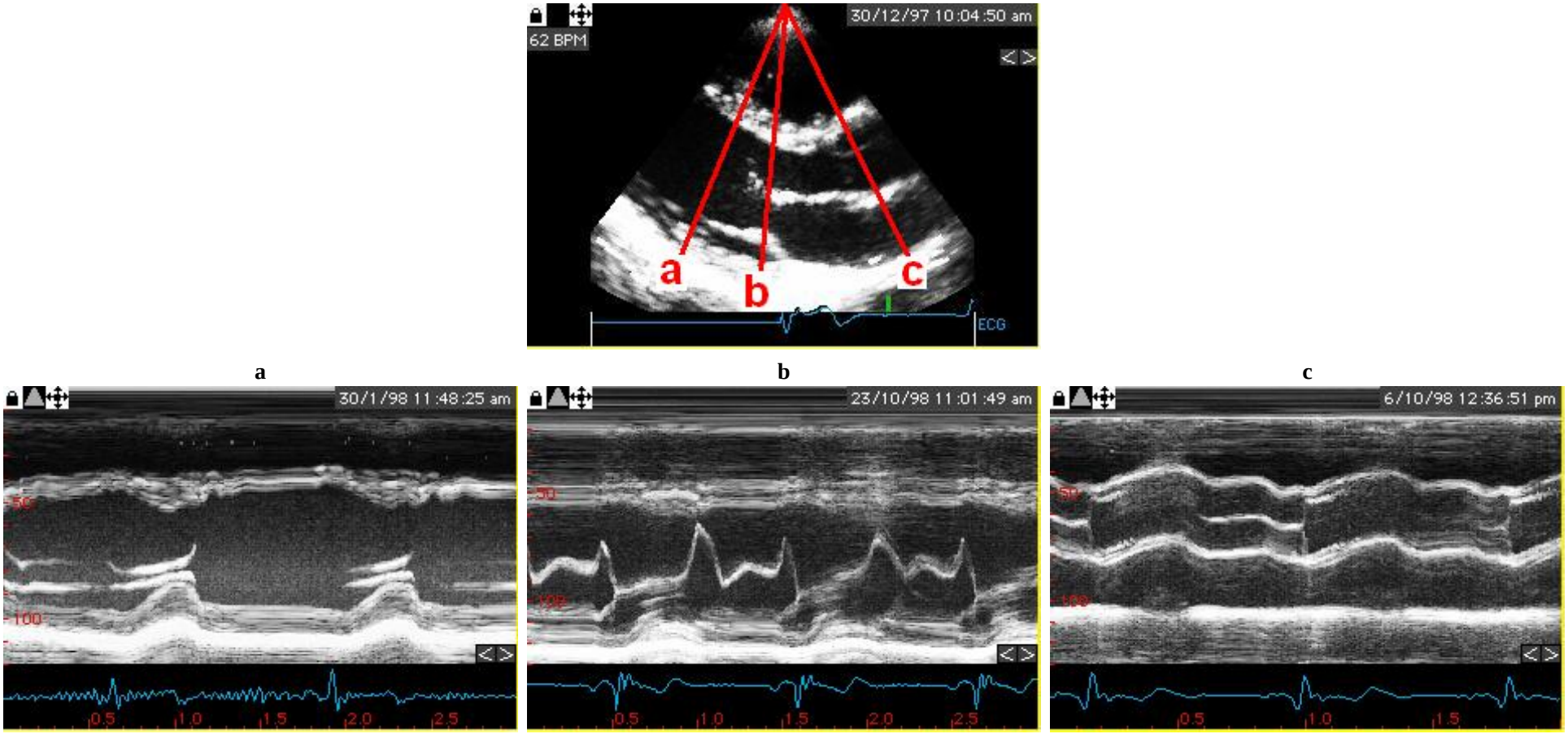


Fig. 4. Typical M-mode images. a from left ventricle, b from the mitral valve and c from the aortic valve as indicated on the 2D long axis image above. Here the amplitude is displayed in white on dark background.

2-dimensional imaging:

If the line of a B-mode sweeps over the object, a 2-dimensional image can be generated.

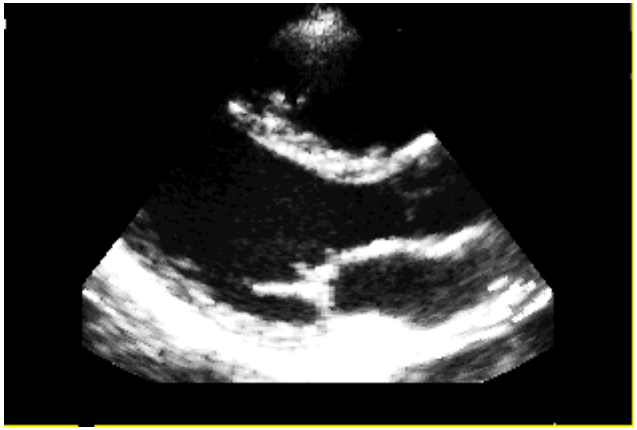


Fig. 5. "D echocardiography. A line is sent out, and as all echoes along the beam are received, the picture along the beam is retained, and a new beam is sent out in the neighbouring region. building up the next line in the image. one full sweep of the beam will then build up a complete image.

This means that as a pulse is sent out, the transducer has to wait for the returning echoes, before a new pulse can be sent out, generating the next line in the image. Thus, the depth of the image, as well as the width (number of lines) limits the possible frame rate.

The building of a 2D picture can be achieved by firing a beam vertically, waiting for the return echoes, maintaining the information and then firing a new line from a neighbouring transducer in a phased **linear array.** The transducer array can be linear, with electronic phased arrays shooting parallel beams in sequence, creating a field that is as wide as the probe length (footprint). A curvilinear array has a curved surface, creating a field in the depth that is wider than the footprint of the probe, making it possible to create a smaller footprint for easier access through small windows. This will result in a wider field in depth, but at the cost of reduced lateral resolution as the scan lines diverge.

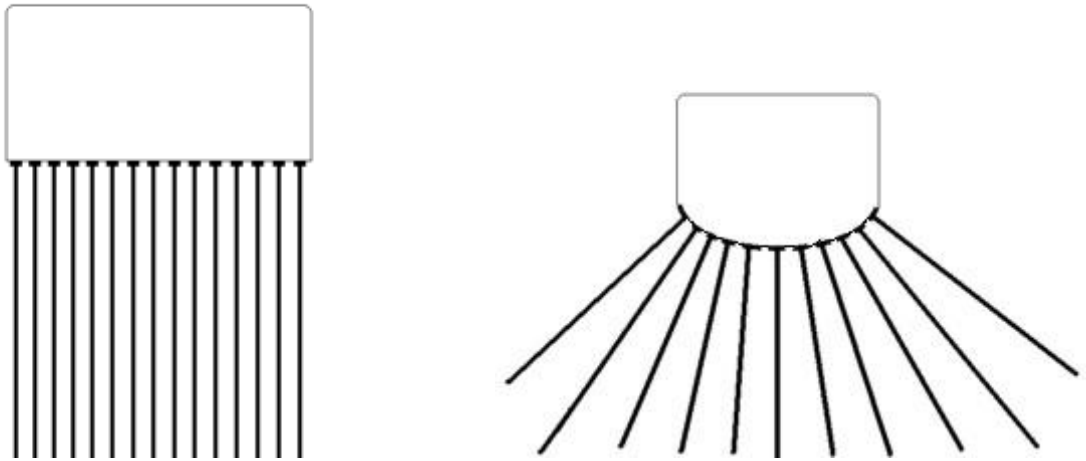


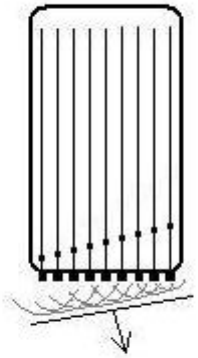
Fig. 6 A. Linear array. **B.** curvilinear array

The linear array gives a large probe surface (footprint) and near field, and a narrow sector. A curvilinear array will also give a large footprint and near field, but with a wide sector.

But in order to achieve a footprint sufficiently small to get access to the heart between the ribs, and with a sufficiently wide far field, the beams has to diverge from virtually the same point. This means that the image has to be generated by a single beam originating from the same point, being deflected in different angles to build a sector image (cfr. fig. 5).

This can be achieved by a single transducer or array sending a single beam that is stepwise rotated, either mechanically or electronically.
A very small footprint can be achieved by a mechanical probe, sending only one beam, but being mechanically rotated by a motor. Finally with a slightly larger footprint, a phased array with electronic focussing and steering, can generate a beam sweeping at an angle similar to the mechanical probe. Beamforming by phased array, also enables focussing of the ultrasound beam as shown. Focussing can also be performed in a mechanical probe, by a concentric arrangement of several ring shaped transducers, an annular array. This will focus the beam in both transverse directions at the same time, as indicated in fig. 7A.

Beamforming



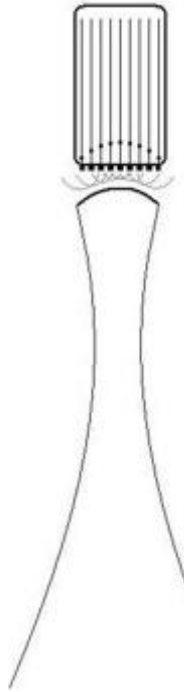


Fig. 7A. Mechanical transducer. The sector is formed by rotating a single transducer or array of transducers mechanically, firing one pulse in each direction and then waiting for the return pulse before rotating the transducer one step. In this beam there is electronic focussing as well, by an annular array.

B. Electronic transducer in a phased array. By stimulating the transducers in a rapid sequence (much more rapidly than in the linear array shown above, the ultrasound will be sent out in an interference pattern. According to Huygens principle, the wavefront will behave as a single beam, thus the beam is formed by all transducers in the array, and the direction is determined by the time sequence. The beam will then sweep stepwise over the sector in the same way as the mechanical transducer in A, sending a beam in one direction at a time.

C. The same principle of phase steering can be applied to make a concave wavefront, resulting in focussing of the beam with its narrowest part a distance from the probe. Combining the steering in B and C will result in a focussed beam that sweeps across the sector, as in A.

Beam focussing:

Focussing is illustrated above. In a mechanical probe, there may be several transducers, arranged in a circular array, focussing the beam in a manner analogous to that shown in fig. 7c. In a circular array, however, the focussing can be done in all directions transverse to the beam direction, i.e. in the imaging plane and transverse to the plane, while a linear array can only focus in one direction, in the imaging plane.

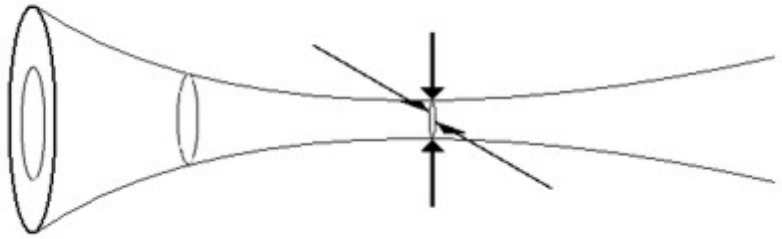
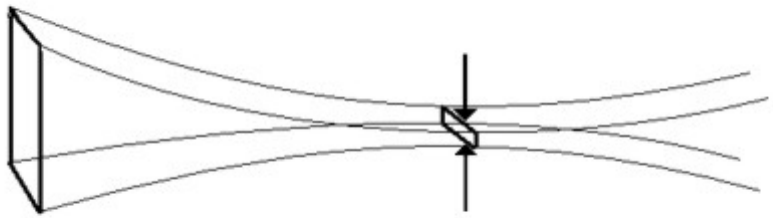


Fig. 8a. Annular focussing in all directions both in plane and transverse to the plane.

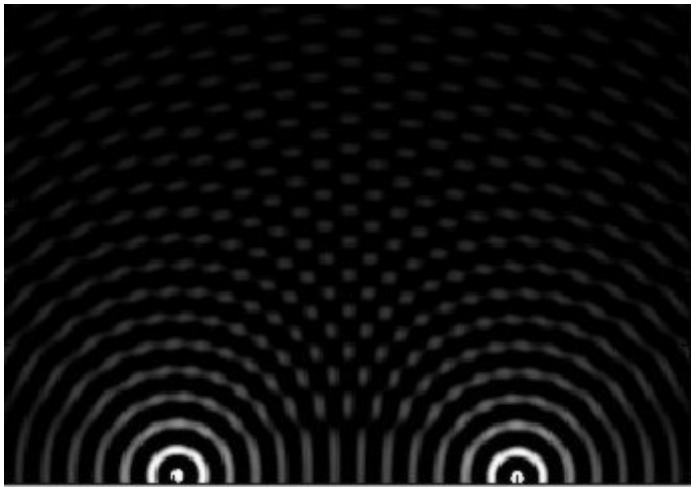


8b. Linear focussing in the imaging plane only.

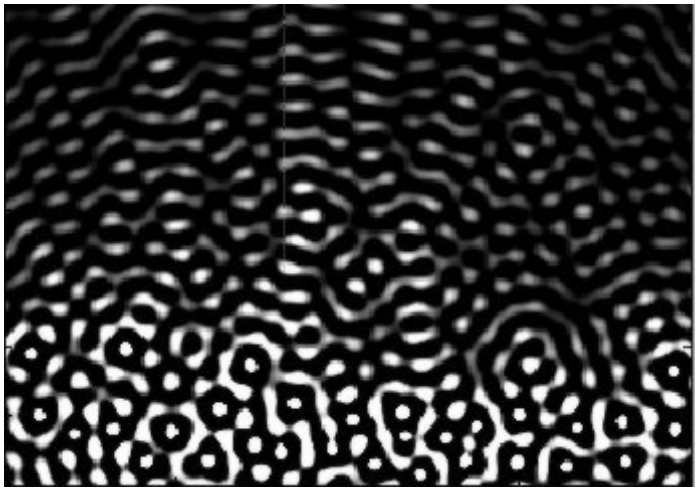
A matrix array, can focus in both directions at the same plane.

Speckle formation:

The gray scale image is seen to consist of a speckled pattern. The pattern is not the actual image of the scatterers in the tiussue itself, but the interference pattern generated by the reflected ultrasound:



Interference pattern. Here is simulated two wave sources or scatterers at the far field (white points). The emmitted or reflected waves are seen to generate a speckle pattern (oval dots) as the amplitude is increased where wave crests cross each other, while the waves are neutralised where a wave crest crosses a though. This can be seen by throwing two stones simultaneously in still water . The speckle pattern can be seen in front of the scatterers, towards the probe.

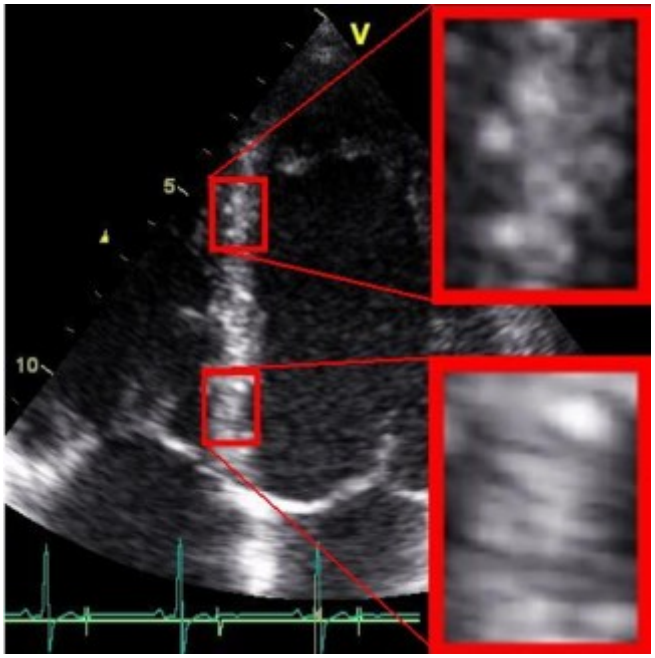


Irregular interference pattern. This is genrated by more scatterers somewhat randomly distributed. The speckle pattern is thus random too. Again there may be a considerable distance between the speckles and the scatterers generating the pattern.

Speckle tracking

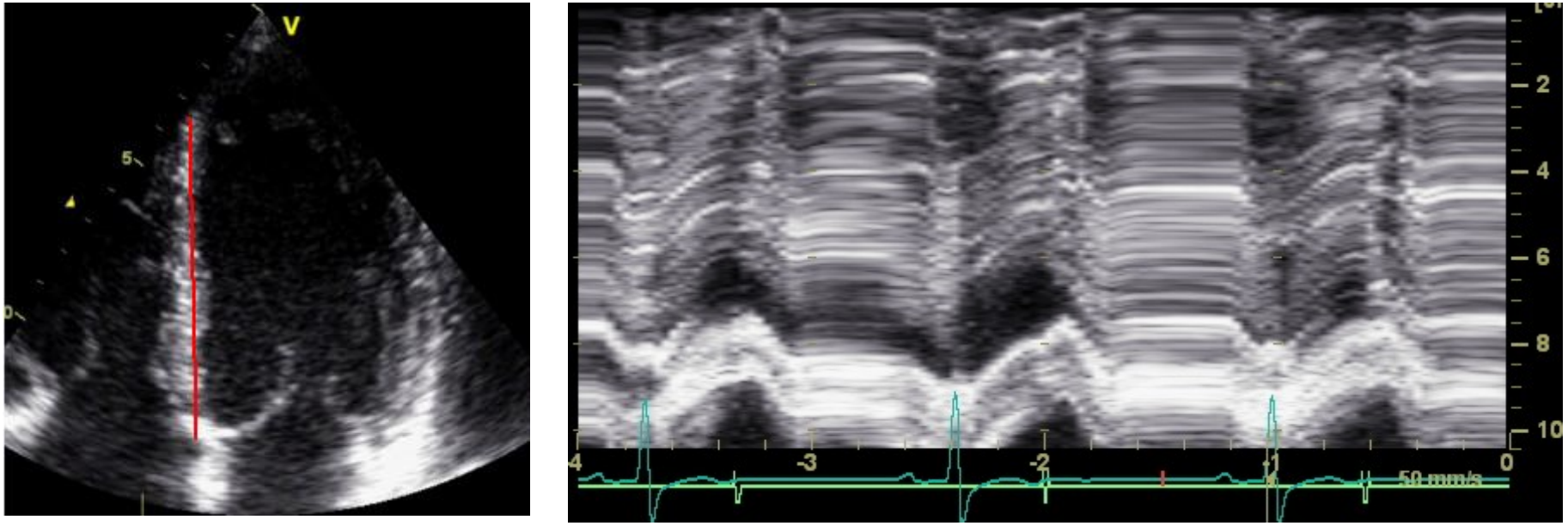
The speckle pattern can be used to track myocardial motion due to two facts about the speckle pattern.

1. The randomness of the speckle pattern ensures that each region of the myocardium has its own unique speckle pattern: that can differentiate a region from other regions.



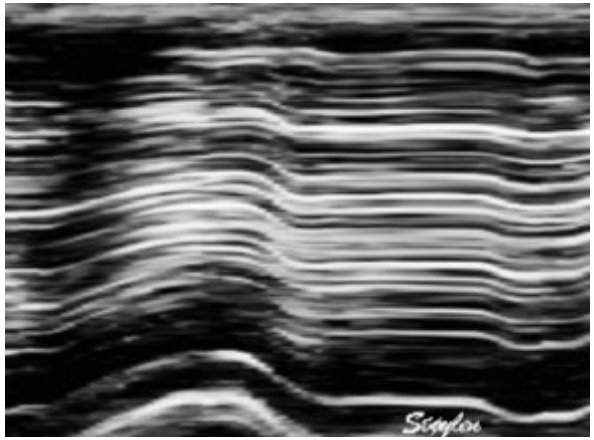
Demonstrating the differnce beween two different regions of the myocardium by their different random speckle pattern.

2. The speckle pattern remains resonably stable, and the speckles follow the myocardial motion. This can be demonstrated by M-mode:

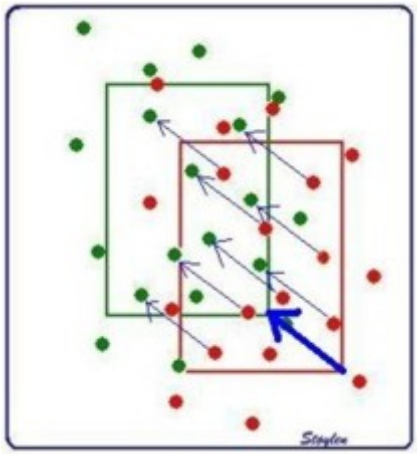


An M-mode along the septum demonstrates how the speckles is shown as motion curves. It is evident that many speckles are only visible during part of the heart cycle, but if the speckle pattern is compared from frame to frame, the changes will be small. The grainy texture of the lines is due to the limited frame rate as the M-mode on the right is reconstructed from the 2D image at the left.

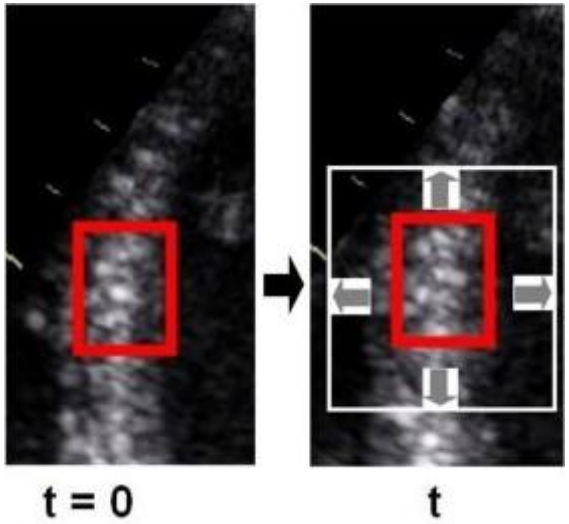
By this, defining a region (kernel) in one frame, this kernel can be identified as region in the next frame with the same size and shape with the most similar speckle pattern, and the motiuon of the kernel can be tracked from frame to frame.



Real time M-mode demonstrates how the speckle pattern follows the myocardial motion.



A search in the next frame identifies the new kernel (green) having a similar speckle pattern (green) as the original kernel in the previous frame (red). The motion of the kernel idntifies the regional tissue motion.



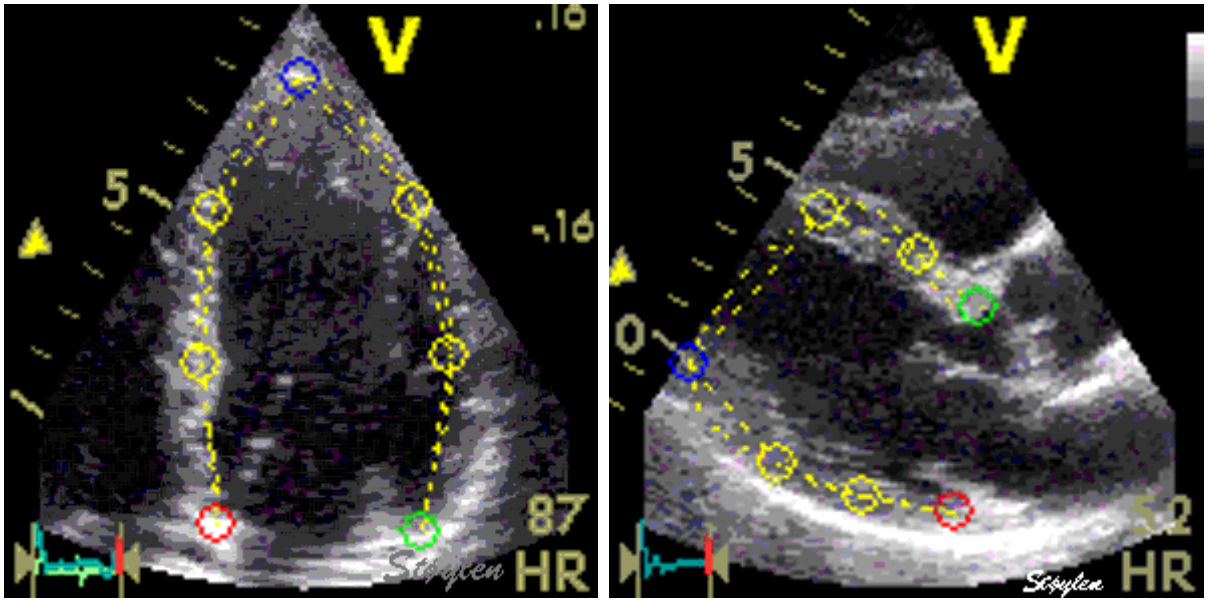
The search algorithm. The kernel is defined in the original frame at t=0 (red square). In the next frame, at t=t, the algorithm defines a search area (white square), and teh search is conducted in all directions for the matching kernel.

The algorithm for this seach is simple, it simply searches for the area with the smallest difference in the total sum of pixel values, the smallest sum of absolute differences (SAD). This has been shown to be as effective as cross correlation (26, 27). However, the speckle pattern will not repeat perfectly. This is due to both true out of plane motion (rotation and torsion relative to apical planes and longitudinal deformation relative to short axis planes) and to small changes in the interference pattern. But the frame to frame change is small, and the approach to recognition is statistical. This means, however, that the search should be done from frame to frame, the changes over longer time intervals will be to great.

Speckle tracking has been validated by ultrasonomicrometry in the longitudinal direction (124) as well as for rotation (125).

More about the mathemathics can be found here.

Tracking is angle independent, and can be used to track in two dimensions:



However, drop outs and reverberations will affect the tracking, and in the lateral direction low lateral resolution will "smear" the speckles in the lateral direction, making tracing less perfect, as can be seen above, where thacking in the inferior wall where lateral resolution is poorer, is less perfect than in the septum where the sector is narrower and lines more dense, giving a better lateral resolution.

Speckle tracking is frame rate sensitive:

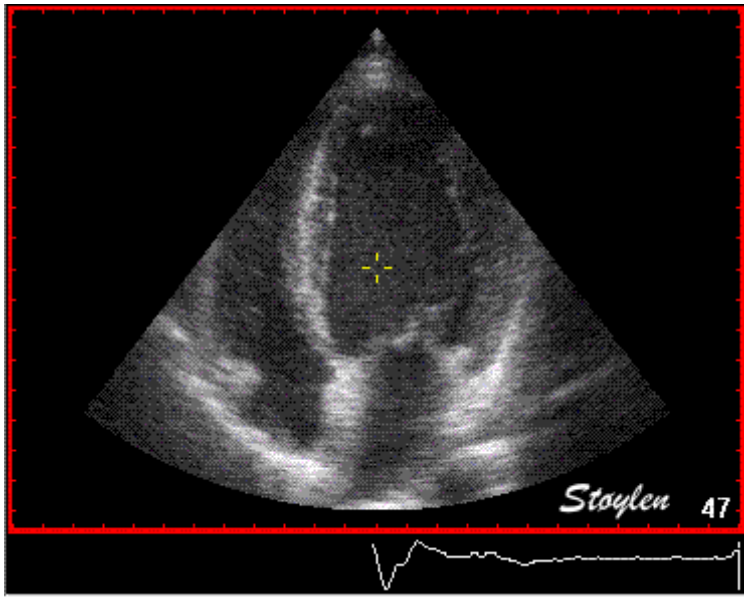
1. Too low frame rate will result in too great changes from frame to frame, resulting in poor tracking. This may also limit the use in high heart rates, as the motion and thus frame to frame change increases relative to the frame rate.
2. Too high frame rate is obtained by reduced lateral resolution, and thus resulting in poorer tracking at least in the transverse direction.

Thus, both too high and too low frame rate may affect speckle tracking adversely. With the present equipment, the optimal frame rate seems to be between 40 - 70 if image quality is good, slightly higher with poorer image quality.

As speckle trecking can track in both transverse and axial directions, with a sufficient number of kernels, deformation can in preinciple be measured in two dimensions, as as dicussed in the section about a new application called "2D strain".

Within its limitations, however, speckle tracking can be used for measuring displacement, velocity, strain and strain rate, as described here.
Speckle tracking can also be combined with tissue Doppler in various combinations for more efficient tracking as described below.

In echocardiography, the beam sweeps over the field 30 - 90 times per second,resulting in the same frame rate. This means that the picture is updated sufficient many times per second to show the motion of the heart as a cine-loop in real time:



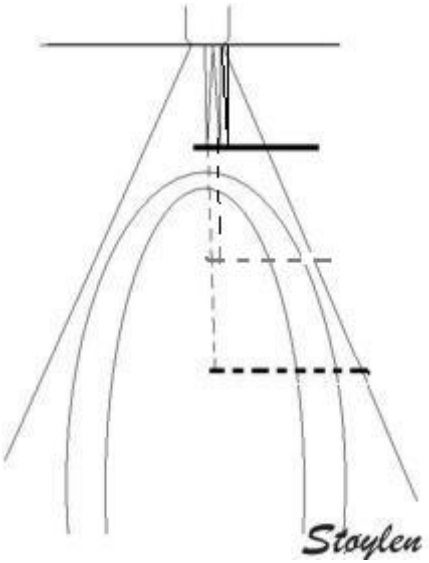
Temporal resolution:

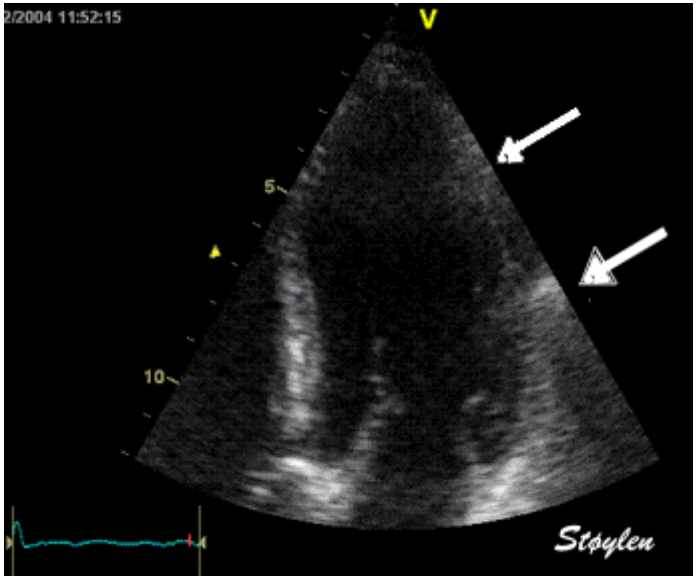
To imagine moving objects, structures such as blood and heart, the frame rate is important, related to the motion speed of the object. This is important not only in 2D moving images, but also in Doppler imaging, see the Nykvist phenomenon. The temporal resolution is limited by the sweep speed of the beam. And the sweep speed is limited by the speed of sound, as the echo from the deepest part of the image has to return before the next pulse is sent out ad a different angle in the neighbouring beam. The sweep speed can be increased by reducing the number of beams in the sector, or by decreasing the sector angle. The first option decreases the lateral resolution, the second decreases the image field, thus basically temporal resolution cannot be increased without a trade off, due to the physical limitations.

Another approach increases the temporal resolution by transmitting wide beams and then receiving the echo from this along several, narrower beams, made possible by the phased array; the MLA technique, 4MLA means that each beam is received along 4 narrower beams. This is utilized especially in Doppler recordings. The problem of this technique is that the energy along the receiving beams are travelling simultaneously, increasing the possibility that energy scattered from a point in one beam is received in the neighbour, leading to a contagion between the beams.

Stationary reverberations:

Stationary reverberations are caused by stationary structures, usually in the chest wall, causing the ultrasound to bounce back and forth between the skin and the structure, increasing the time before the echo returns and giving rise to an apparent stationary structure deeper down.

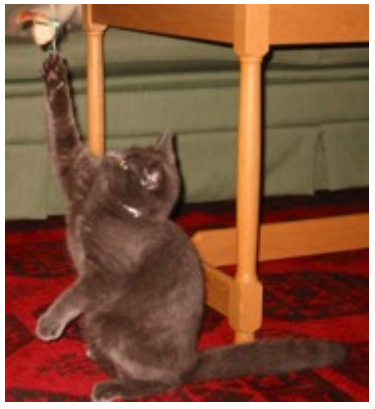




Left, the principle of stationary reverberations, showing how the ultrasound pulse bounces twice between a stationary structure in the body wall and the surface (thick line), causing the reflected pulse to arrive the transducer three times as late as if it hadn't bounced (dotted, thick line). This is then interpreted as the reflector being located three times it's true depth. Right, the appearance in grey scale image, see the stationary echo in the lateral wall (arrows).

The Doppler effect

Christian Andreas Doppler (1803 - 1853)



My cat Doppler (2004 -

The Doppler effect was discovered by Christian Andreas Doppler (1803 - 1853), and shows how the frequency of an emitted wave changes with the velocity of the emitter or

observer. The theory was presented in the royal Bohemian society of Science in 25th of May1842 (5 listeners at the occasion!), and published in 1843 (119). The premises for his theoretical work was faulty, as he built his theory on the work of James Bradley who erroneously attributed the apparent motion of stars against the background (parallax effect) to the velocity of the earth in its orbit (instead of the effect of Earth's position in orbit on the angle of observation). Further, Doppler attributed the differences in colour of different stars to be due to the Doppler effect, assuming all stars to be white. Finally, he theoretised over the effect of the motion of double stars that rotate around each other, assuming a Doppler effect from the motion. The changes in wavelength from the Doppler effect, however, is too small to be observed.

However, Doppler did a theoretical derivation of the effect of the motion of the source or observer on the perceived wavelength from the premises of a constant propagation velocity of the waves in the medium, and this is entirely correct, valid both for sound waves and electromagnetic radiation of all kinds. The basis for the Doppler effect is that the propagation velocity of the waves in a medium is constant, so the waves propagates with the same velocity in all directions, and thus there is no addition of the velocity of the waves and the velocity of the source. Thus, as the source moves in the direction of the propagation of the waves, this does not increase the propagation velocity of the waves, but instead increases the frequency.The original derivation of the Doppler principle as well as the extension to reflected waves is explained in more detail here. As a work of theoretical physics, it is thus extremely important. In addition, it has become of practical importance, as the basis for the astronomical measurement of the velocity of galaxies by the red shift of the spectral lines, in Doppler radar, Doppler laser and Doppler ultrasound.

The theory was experimentally validated by the Dutchman Christoph Hendrik Diderik Buys Ballot (120), with the Doppler effect on sound waves, who placed musicians along a railway line and on a flatbed truck, all blowing the same note, and observed by subjects with absolute pitch, who observed the tones being a half note higher when the train was approaching as compared to the stationary musicians, and a half note lower as the train receded.

(This can be observed in everyday phenomena such as the sound of f.i. an ambulance siren, the pitch (frequency) is higher when the ambulance is coming towards the observer, hanging as it passes, and lower as it goes away.

This is illustrated below:

Stoylen

The Doppler effect. As the velocity of sound in air (or any other medium) is constant, the sound wave will propagate outwards in all directions with the same velocity, with the centre at the point where it was emitted. As the engine moves, the next sound wave is emitted from a point further forward, i.e. with the centre a little further forward. Thus the distance between the wave crests is decreased in the direction of the motion, and increased in the opposite direction. As the distance between the wave crests is equal to the wavelength, wavelength decreases (i.e. sound frequency increases) in front of the engine, and increases (sound frequency decreases) behind it. This effect can be heard, as the pitch of the train whistle is higher coming towards a listener than moving away, changing as it passes. The effect on the pitch of the train whistle was published directly, but later than Doppler and Buys Ballot.

If the sound source is stationary, the effect on moving observer is similar. The train will meet the wave crest with shorter intervals, as the train moves into the incoming sound. In ultrasound, the wave is sent from a stationary transducer, the moving blood or muscle is firstly moving towards the transducer and then following the reflected wave towards the transducer, thus the Doppler shift is approximately twice as great. In the case of reflected ultrasound, the Doppler shift is:

$$f_D = 2 f_0 \frac{v}{c} \cos(\alpha)$$

where α is the angle between the direction of the motion and the ultrasound beam.

Thus, in the case of reflected ultrasound, the velocity of blood or tissue can be measured by the Doppler shift of the reflected ultrasound:

Basically, the Doppler effect can be used to measure blood and tissue velocities from the Doppler shift of reflected ultrasound:

$$v = \frac{f_D \cdot c}{2 f_0 \cdot \cos(\alpha)}$$

where v is the blood or tissue velocity, c is the sound velocity in tissue, f_0 is the transmitted frequency, f_D is the Doppler shift of reflected ultrasound and α is the insonation angle, between the ultrasound beam and the direction of motion (velocity vector).

Pulsed and continuous wave Doppler:

Doppler pulses can either be used as a pulsed Doppler, where a pulse is sent out, and the frequency shift in the reflected pulse is measured at a certain time. This will correspond to a certain depth, i.e. velocity is measured at a specific depth, which can be adjusted. The width is the same as the beam width, and the length of the sample volume is equal to the length of the pulse. The same transducer is used both for transmitting and receiving.

A problem in pulsed Doppler is that the Doppler shift is very small compared to the ultrasound frequency. This makes it problematic to estimate the Doppler shift from a single pulse, without increasing the pulse length too far. A velocity of 100 cm/s with a ultrasound frequency of 3.5 MHz results in a maximum Doppler shift of 2.3 KHz. The solution to this problem is shooting multiple pulses in the same direction and produce a new signal with one sample from each pulse, the Doppler curve from this signal will be a new curve with the frequency equal to the Doppler shift. (This means that a full package of pulses is considered one pulse in the

The pulsed modus results in a practical limit on the maximum velocity that can be measured. In order to measure velocity at a certain depth, the next pulse cannot be sent out before the signal is returned. The Doppler shift is thus sampled once for every pulse that is transmitted, and the sampling frequency is thus equal to the pulse repetition frequency (PRF). Frequency aliasing occurs at a Doppler shift that is equal to half of the PRF.

$$f_D = ½ * PRF$$

This is illustrated below with an analogy:

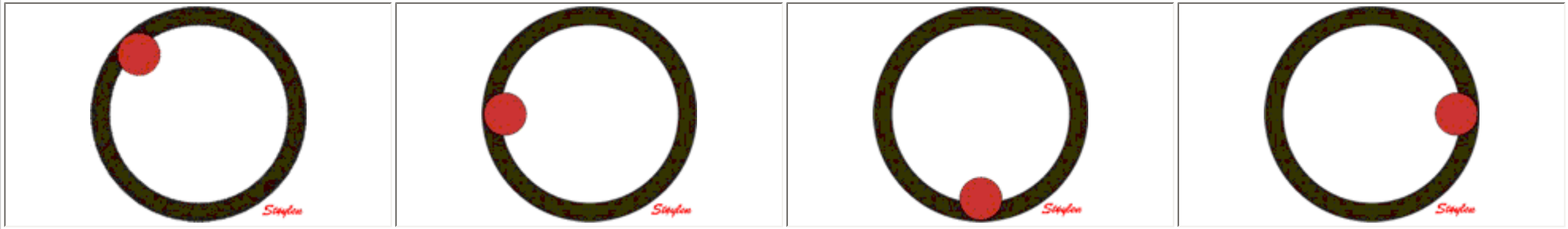
The Nykvist phenomenon.

The Nykvist phenomenon (121) is an effect of the relation between the sampling frequency and the observed velocity. If you sample at a certain frequency, the direction of the motion becomes ambiguous, more frequent sampling will give the correct direction, less frequent sampling results in an apparent motion in the opposite direction. This can be observed with a stroboscopic light, for instance illuminating the flow of water, or with old fashioned wagon wheels in old moves which often seem to revolve slowly backwards when the wagon moves forwards.

This is illustrated below.

Constant rotation velocity, decreasing sampling frequency:

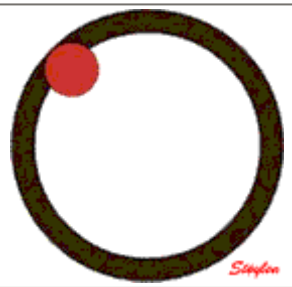
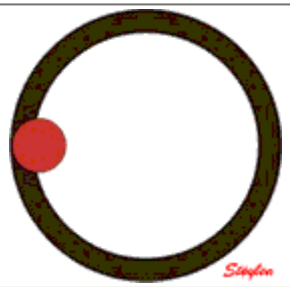
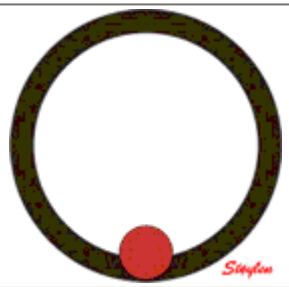
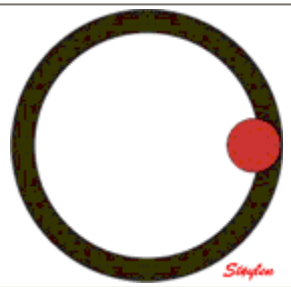
The easiest is to show how reducing the sampling frequency affects the apparent motion. All circles rotate with the same rotation velocity clockwise. The sampling frequency is reduced from left to right. It can be seen that the red dots is at the same positions when they are seen to move.



a: 8:1 8 samples per rotation, the red point is seen in eight positions during the rotation.	**b: 4:1** 4 samples per rotation, the red point is seen to rotate just as fast, but is only seen in four positions	**c: 2:1** 2 samples per rotation, i.e. the sampling frequency is exactly half the rotation frequency. Here, the red dot is only seen in two positions, and it is impossible to decide which way it is rotating. This is the Nykvist limit.	**d: 1.5:1** 1.5 samples per rotation,or one sample per three quarter rotation, making it seem that the red dot is rotating counter clockwise.

Constant sampling frequency, increasing rotation velocity

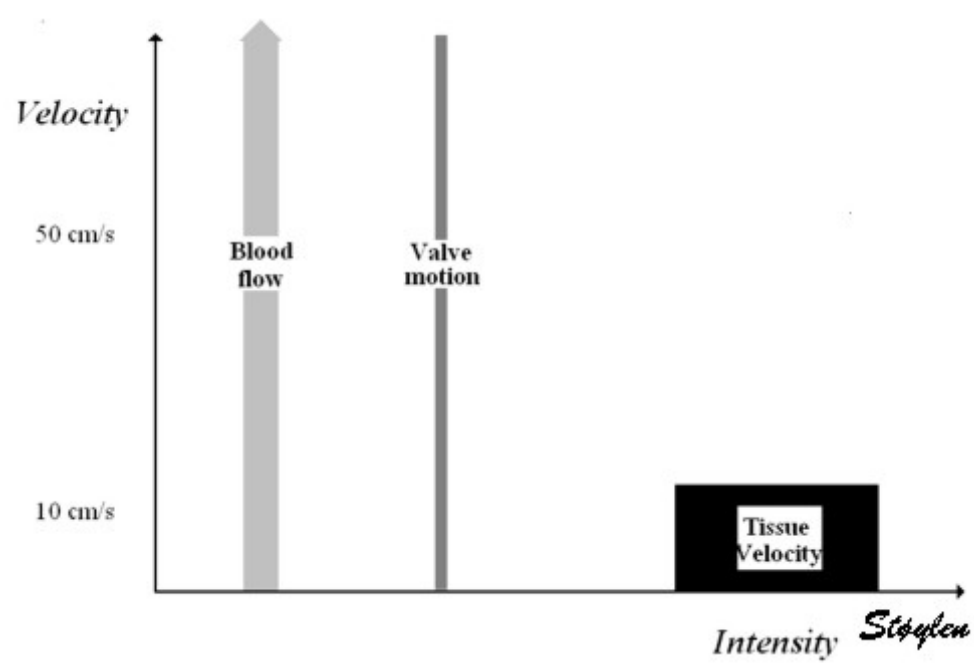
The same principle applies when there is a fixed sampling frequency, but increasing rotational velocity. In the images below, the frames are seen to shift simultaneously, but the positions of the red dots are different due to the different rotational velocity.

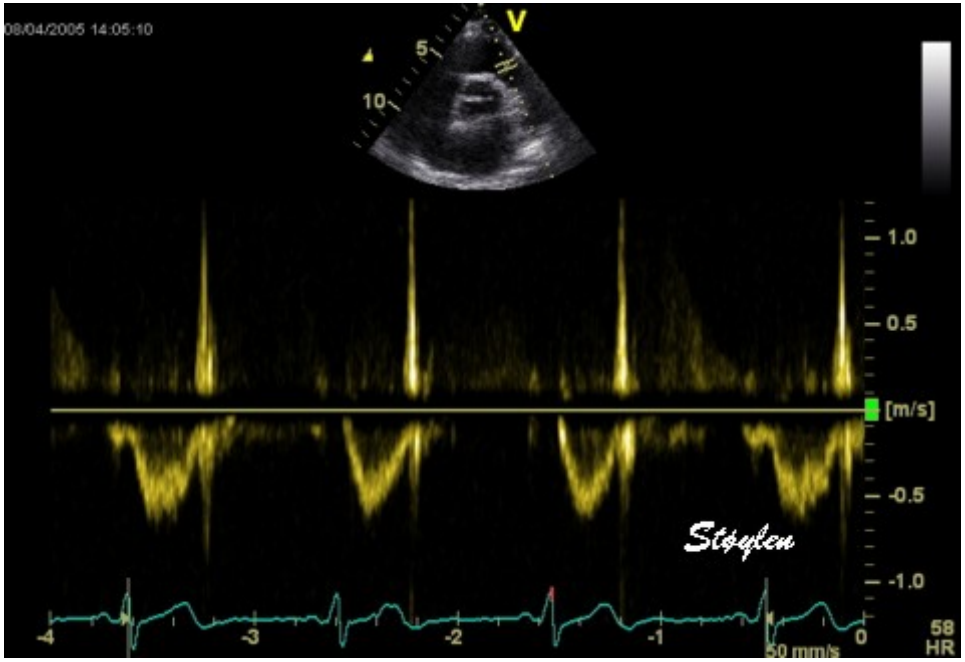
			
a: 1:8 One rotation per 8 samples. The sampling catches the red dot in 8 positions during one rotation.	**b: 1:4** Rotation velocity twice that i a; one rotation per four samples, the sampling catches the red dot only in four positions during one rotation.	**c: 1:2** Rotation velocity four times a; one rotation per two samples, this catches the red dot in only two positions, giving directional ambiguity as above.	**d: 1:1,5** Rotation velocity six times a; one rotation per 1,5 samples, or 3/4 rotation per sample, giving an apparent counter clockwise rotation.

Sampling from increasing depth will increase the time for the pulse returning, thus increasing the sampling interval and decrease the sampling frequency. The Nykvist limit thus decreases with depth. This means that pulsed Doppler has depth resolution, but this leads to a limit to the velocities that can be measured.

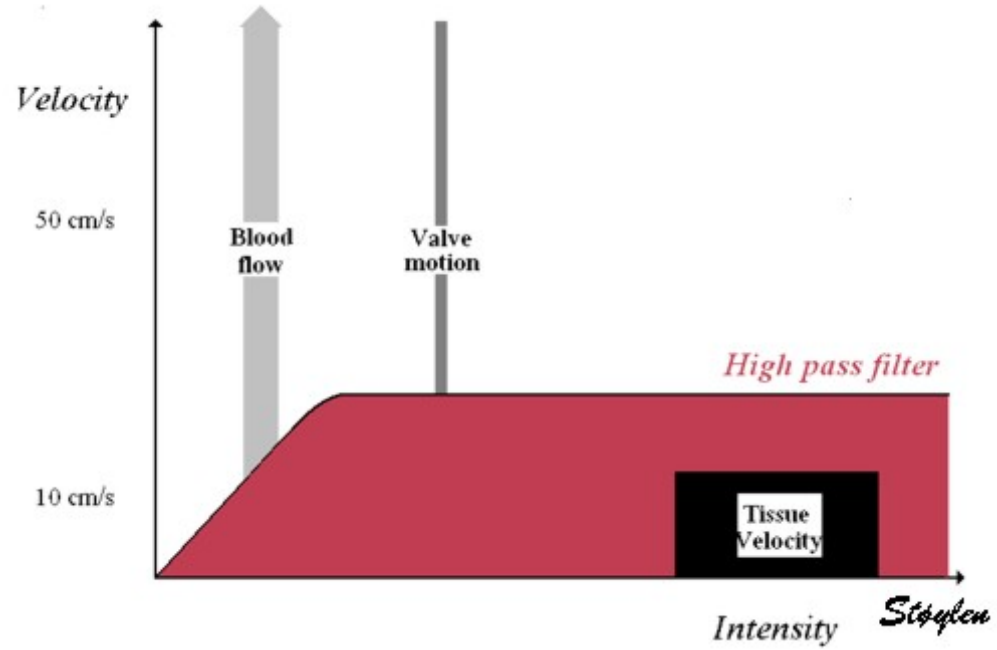
Tissue Doppler.

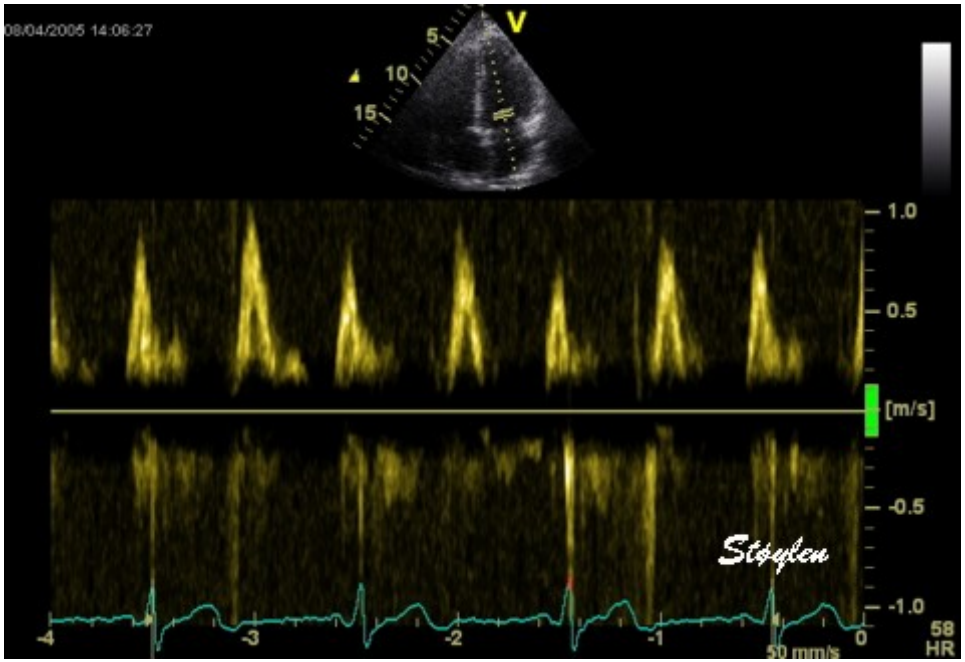
The Doppler principle can be used both for blood flow and Tissue velocities. The main principle is that blood has high velocity (Typically above 50 cm/s, although also all velocities down to zero), but low density, resulting in low intensity (amplitude) reflected signals. Tissue has high density, resulting in high intensity signals, but low velocity (typically below 20 cm/s). The difference in the applications used for the two sets of signals is mainly differences in filtering, applying a high pass filter in Doppler flow, and low pass filter in tissue Doppler (Although the latter is not absolutely necessary).



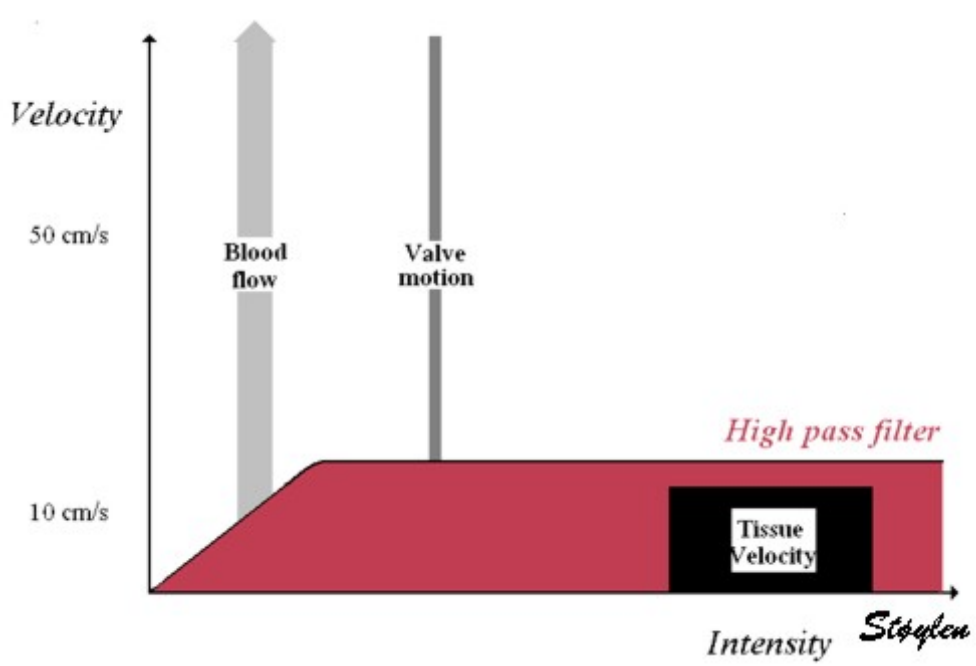


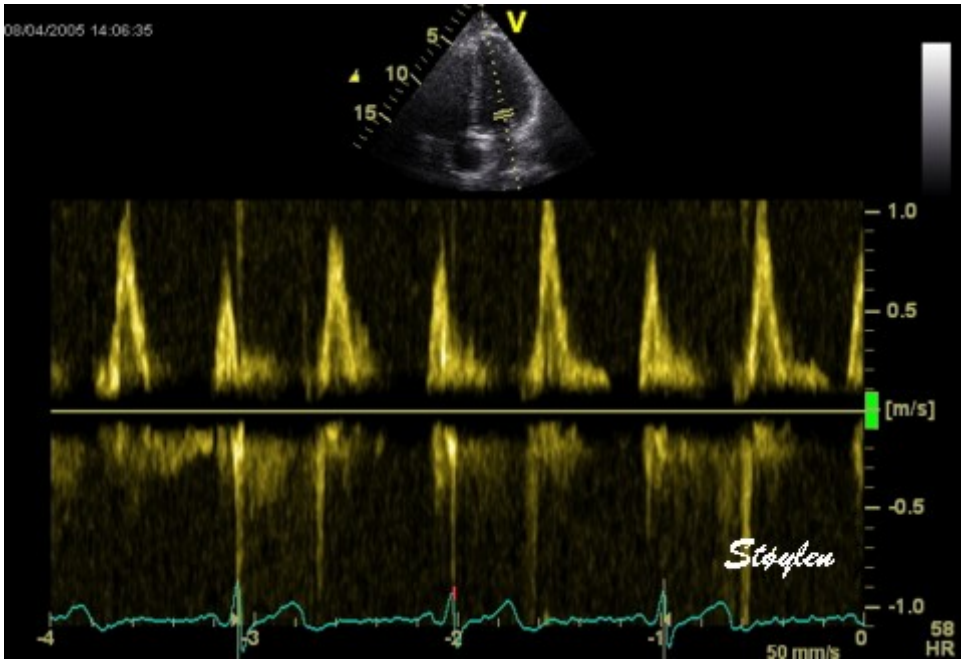
The diagram to the left shows the placement of flow and tissue signals on this intensity (amplitude) diagram. The flow signals are low intensity but mostly high velocity, while the tissue is exclusively low velocity, high intensity. The heart valves, however, are solid structures which moves with the velocity of the passing blood, resulting in high intensity signals giving a saturation of the Doppler spectrum. A typical flow curve from the right ventricular outflow tract is shown to the left, with the valve click.



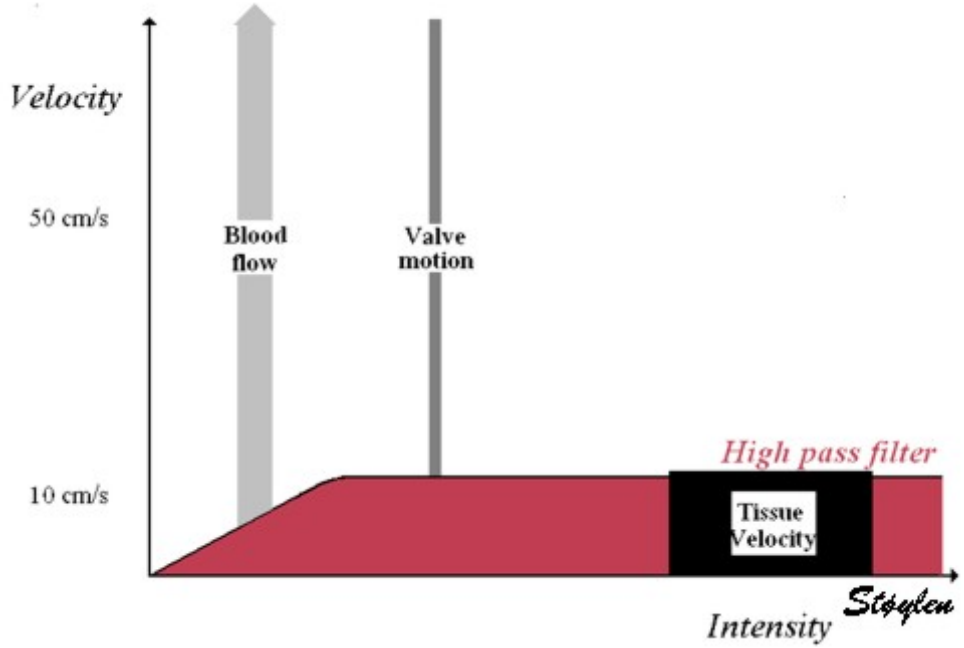


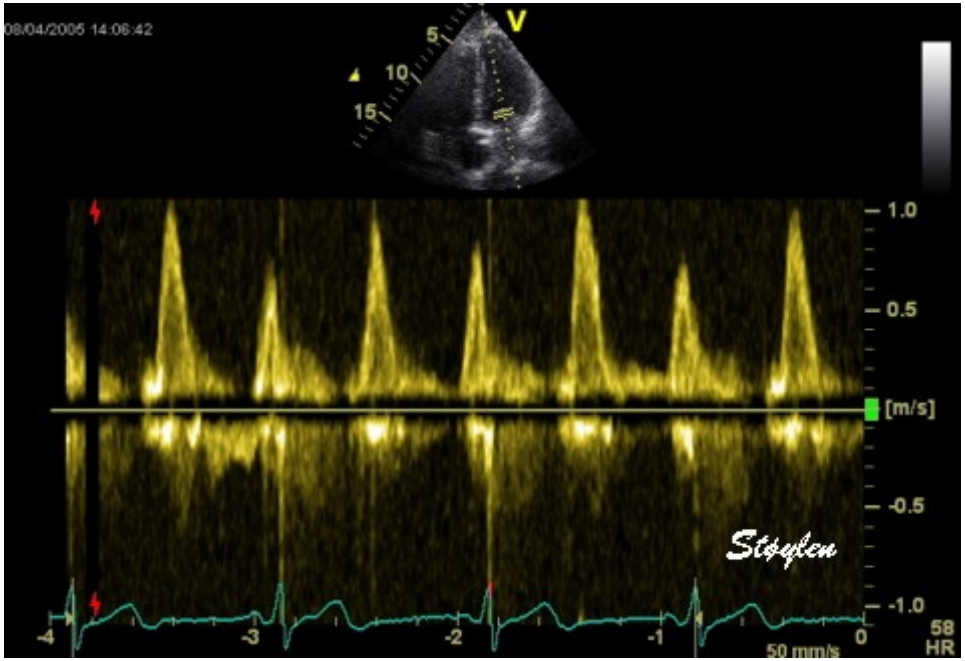
Application of a high pass filter (low velocity reject) shown schematically to the left and in practice applied to a mitral flow curve to the right. The setting rejects velocities at blood intensities below 15 - 20 cm/s, which is too high for normal flow velocities as in this instance, although may often be useful in continuous wave Doppler recordings of high velocities in jets.



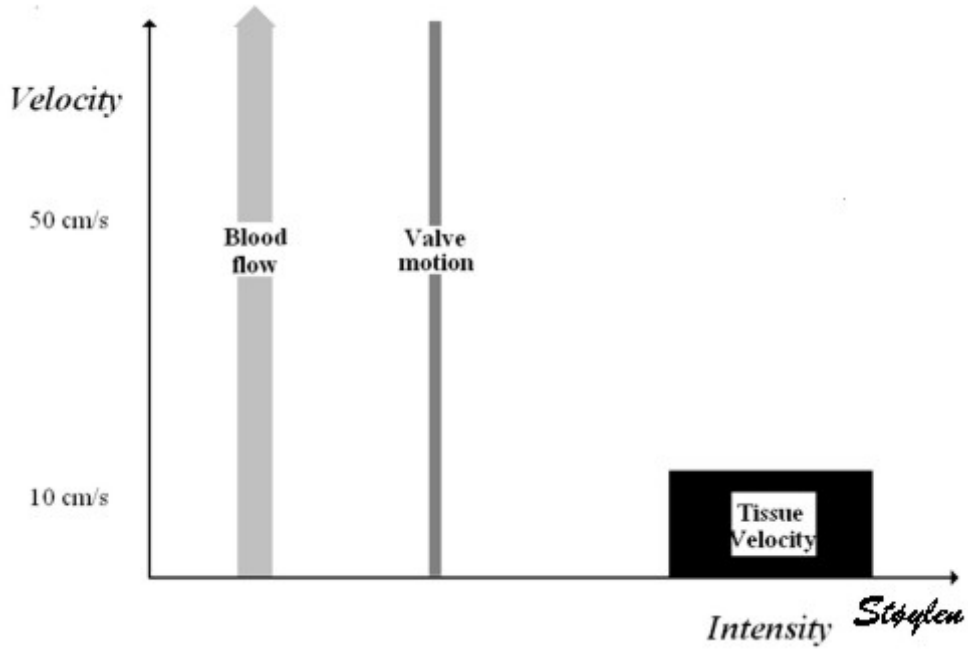


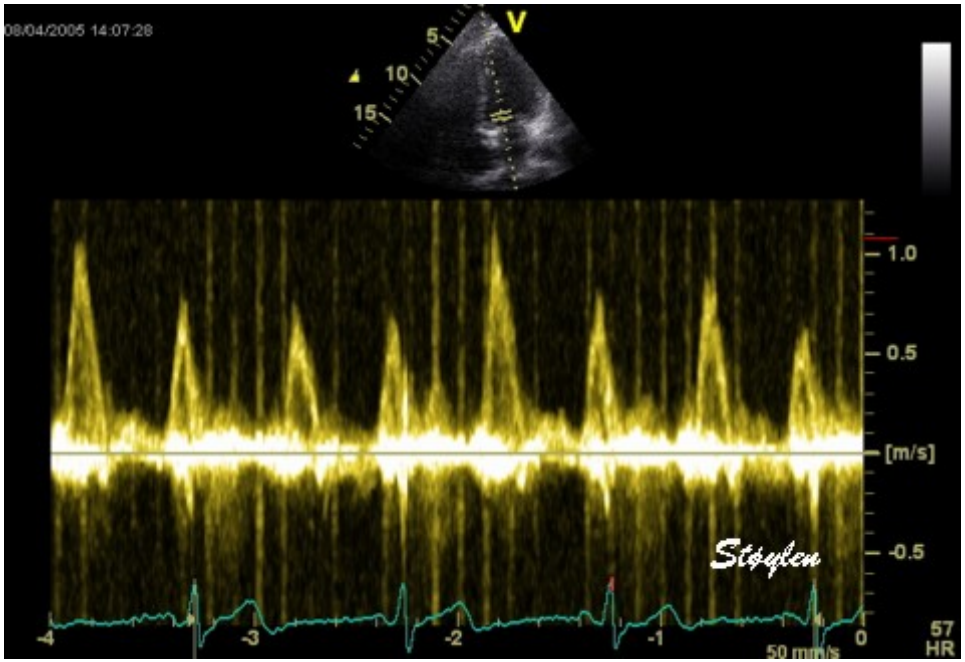
The filter is adjustable and is here reduced to 10 cm/s



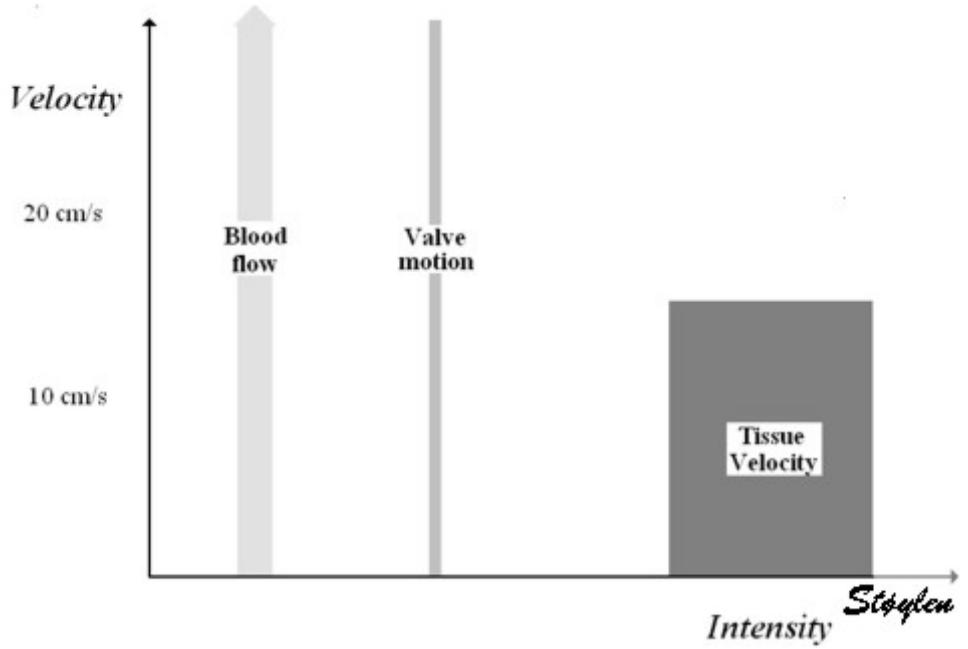


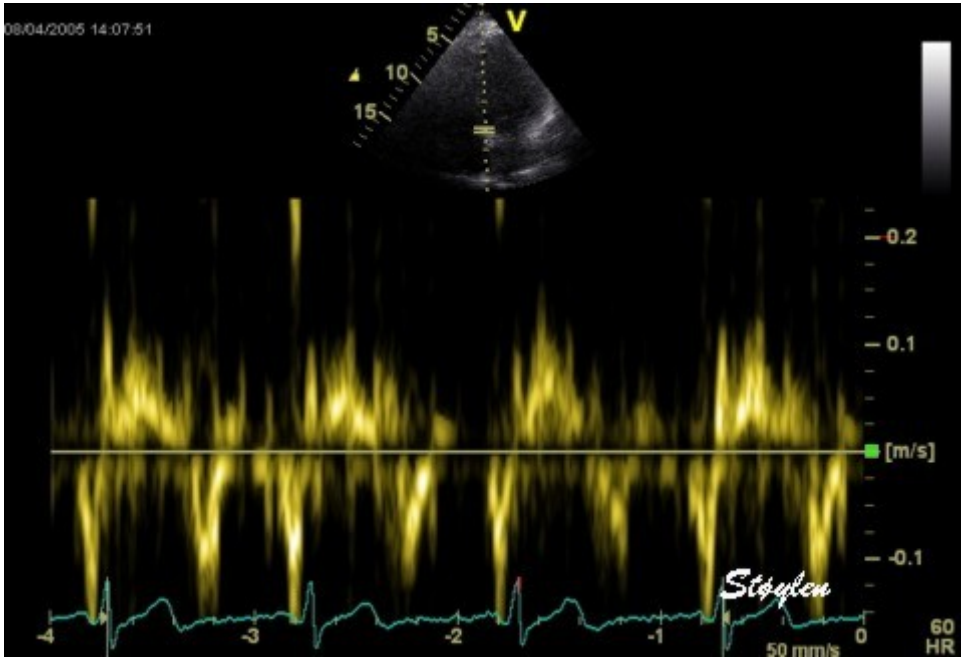
Further reduction in the filter below 10 cm/s results in high intensity signals becoming visible, especially in early diastole. This is tissue signals from the mitral ring.



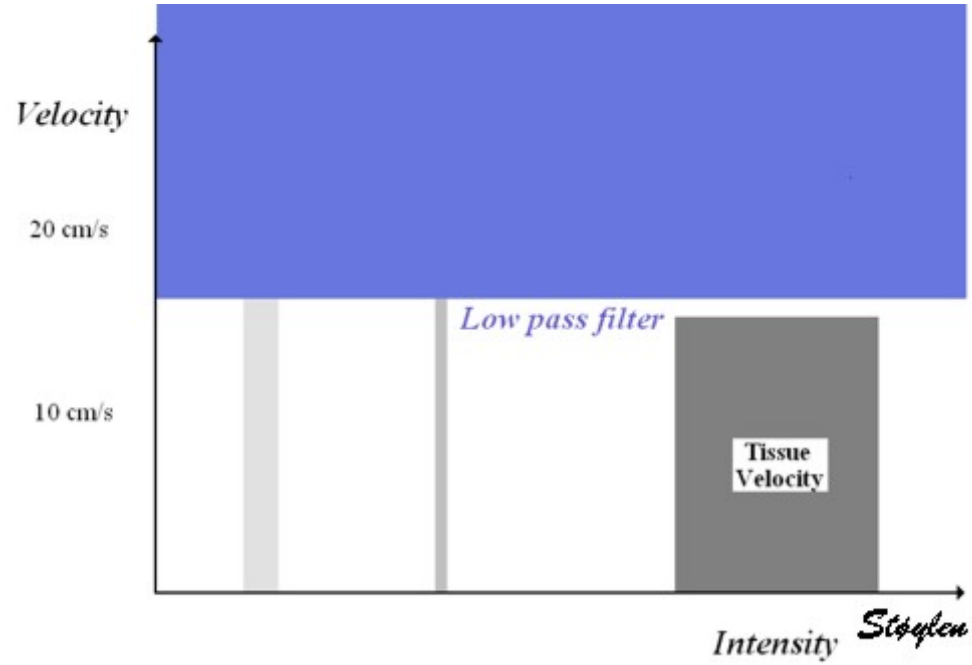


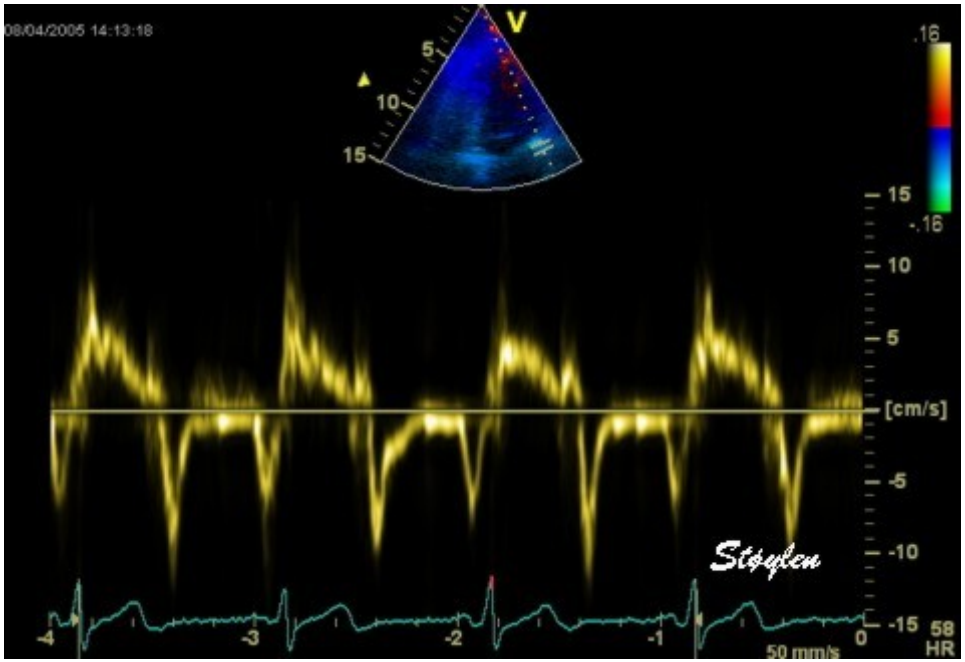
Fully removing the filter results in a dense band of high intensity tissue signals around the baseline.





Decreasing the scale and gain, (and placing the sample volume in the mitral ring), discloses the tissue velocity curve of the ring, still taken with an ordinary Doppler.





All modern ultrasound machines today has separate applications for tissue Doppler which optimises the signal for this purpose, among other things by applying a low pass filter that removes most of the flow velocities.

Tracking by combined speckle tracking and tissue Doppler

Tracking tissue motion by speckle tracking is described above. Modern ultrasound equipment has the capability of acquiring second harmonic greyscale images with an acceptable frame rate of 40 - 50 FPS and good lateral resolution simultaneously with tissue Doppler data. This opens the possibility of tracking along the ultrasound beam by tissue Doppler, simply by calculating the displacement from one frame to next from the velocity and frame interval, while tracking transverse to the ultrasound beam can be done by speckle tracking (124).

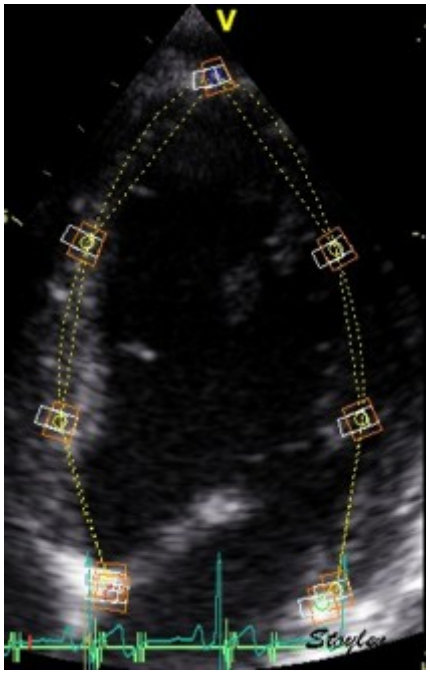


Image showing the speckle tracking from both modalities. The kernels are shown as the small, round, yellow circles. The longitudinal search area along the ultrasound beam by tissue Doppler is shown in red. The lateral search area by speckle tracking is shown in white.

This has several advantages:

1. It increases computational speed, as the SAD is calculated over a far smaller area, especially as the longitudinal motion is greater than the transverse.
2. If the method is used to compute longitudinal velocities or strain rate, the longitudinal tgracking is done with the high sampling frequency of tissue Doppler, thus reducing the possibility of undersampling.
3. It utilises the full dataset inherent in the combined image. **It is not reasonable to assume that discarding part of the information in the acquisition results in better data.**

This method can be used in different ways to analyse strain rate imaging (127) as decribed here and has even been shown to be clinically useful in stress echo (128).

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